NON-INVASIVE SYSTEM FOR UBIQUITOUS PHYSIOLOGICAL HOME MONITORING

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Abstract

In this paper, we propose a non-invasive system of measurement for home care. The system consists in three parts: the sensors, the hardware and the software. The piezoelectric sensor measure the movements and vibration coming from the body (pulse and respiration). The hardware amplifies and filters the signal coming from the piezoelectric sensor before being acquired by a microcontroller and then sent to a computer. The software receives the data and separates the signals coming from the different sensors. Suitable sensors are used for evaluating the ambient conditions, where sound, light intensity, temperature and humidity are measured. The signal captured from the piezoelectric sensor is validated using Pearson correlation coefficient with an ECG synchronized signal. The light and sound sensors are calibrated, using standard equipment: a lux meter (lux) and a sound meter (dB). The proposed unobtrusive system is a convenient tool for ubiquitous health monitoring, simple to implement and able to provide relevant information without interfering with people's daily activities.

Keywords

home care, ubiquitous monitoring, ballistocardiogram (BCG), unobtrusive

Introduction

Home care has been developed as a modern way to take care of patients. It is comfortable for the patient, promotes better life quality and reduces workload in hospitals [1]. However, regular monitoring devices can be cumbersome and uncomfortable, particularly for chronic patients. This paper presents a proposal for a new non-invasive system to measure physiological variables, particularly heart rate, respiration rate and movements of the body. Also, the system monitors the environment, measuring sound, light, humidity and temperature of the room to complement the physiological data.

Body Movements

Pulse and respiration movements are sufficient to determine heart rate and respiratory rate. From both measurements it is possible to detect arrhythmias, bradycardia, tachycardia and apneas [2]. It is also possible to obtain objective indexes of how people sleep from number of apneas, arrhythmias, number of body movements, sleep cycles and time in bed [3].

Pulse

The heart pumps the blood to distribute it to the entire body. The force applied by the heart to pump the blood through the arteries generates an opposite reaction movement of the body. The measure of that movement is the ballistocardiogram (BCG) (Fig. 1). BCG is a graphical representation of the pulse generated when the heart pumps. The key fiduciary point are I, J, K which

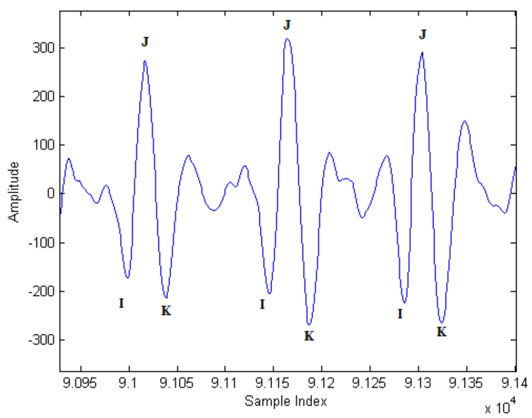

Fig. 1: Typical BCG waveform showing I, J, K points.

correspond to the mechanical response to the ejection of blood from the ventricles.

Respiration

Non-invasive respiration monitoring relies on detecting movements of the ribcage, involving mainly the diaphragm, intercostal and abdominal muscles. The respiration waveform is used for respiration rate calculation and apnea detection.

General System

The system is sub-divided in three parts: the sensors, the hardware and the software.

Sensors

Several authors use piezoelectric sensors for non-invasive measurements [4], [5]. For this system, there are 2 candidates, EMFi (Electro Mechanical Film) and PVDF (polyvinylidenefluoride) sensors. The EMFi is composed by gas bubbles, and when a force is applied to it, a charge is generated between the electrodes of the film. The PVDF sensor is similar to EMFi, responding to variation of force, but it is made with dipoles. Without forces, the net charge is zero, and when a force is applied, the dipoles change their orientation and generate a charge. Both sensors have enough sensitivity to capture movements of the body. When the blood is pumped by the heart and the respiration produces abdominal or thoracic movements, those pressure changes are captured by these sensors. The sensors are usually placed in the seat and the backrest, to detect pulse and respiration. Our preliminary tests concluded that EMFi sensors have better gain, providing a better Signal to Noise Ratio.

To measure the sound intensity, a LMV321 sound detector is used, whose output voltage is related to the sound intensity. Envelope acquisition mode (only peak to peak values are reported) is used to ensure patient privacy. The sensor has no means to translate from voltage to sound pressure, so a calibration was needed. Fig. 2 shows different values captured at 200 S/s using a 100 point moving average filter as the volume is increased. Fig. 3 shows the final calibration curve using a Sound Meter.

The ambient light sensor TEMPT6000 is used to measure the light intensity in the room. The analog voltage output is proportional to the light intensity. The microcontroller captures this signal using 200 S/s.

A temperature and humidity sensor SHT15 is was used. It is pre-calibrated from factory. It has 14 bits of resolution for temperature, and 12 bits of resolution for relative humidity. An I2C communication is used to measure the temperature and humidity ambient.

Hardware

The hardware has three parts: the amplifier, the filters and the acquisition. To amplify the signal coming from the sensor, a charge amplifier is used, then filtered by a low pass filter with cutoff frequency of 30 Hz (Fig. 4), and is captured by a microcontroller ATMega64 using its Analog to Digital Converter, with a sample rate of 200 S/s. After that, the data is sent to the computer using wireless RS232 serial communication, through XBee series 2 modules.

Software

The raw signal acquired from the sensors contains respiration, BCG and body movements. It is necessary to separate them in order to calculate the respiratory rate, apneas, heart rate and numbers of movements. Some authors use digital filters to separate the respiration signal and cardiac signal, because they have different frequency bands. The respiratory activity has a frequency band between 0.1 and 0.5 Hertz, and cardiac activity has a band between 1 and 4 Hertz, so FIR bandpass filters could be used to extract the BCG signal and the respiratory signal. However, since the bands are so close, it is difficult to implement such bandpass filters. A better solution is using Discrete Wavelet

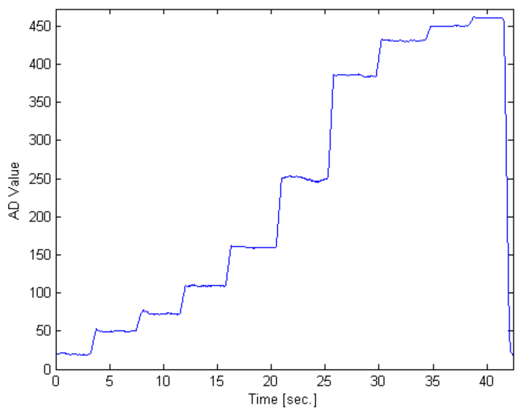

Fig. 2: Sound sensor output during sound test using 800 Hz tone and different intensities.

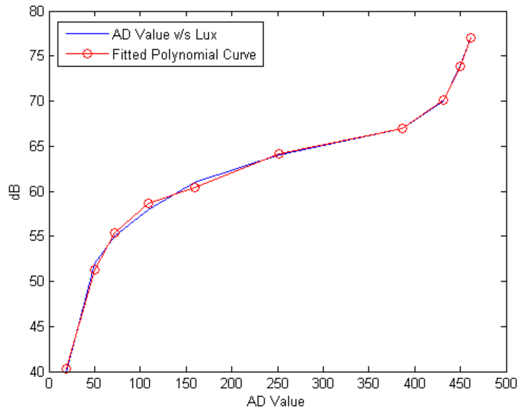

Fig. 3: Polynomial relation between AD conversion and sound intensity (dB).

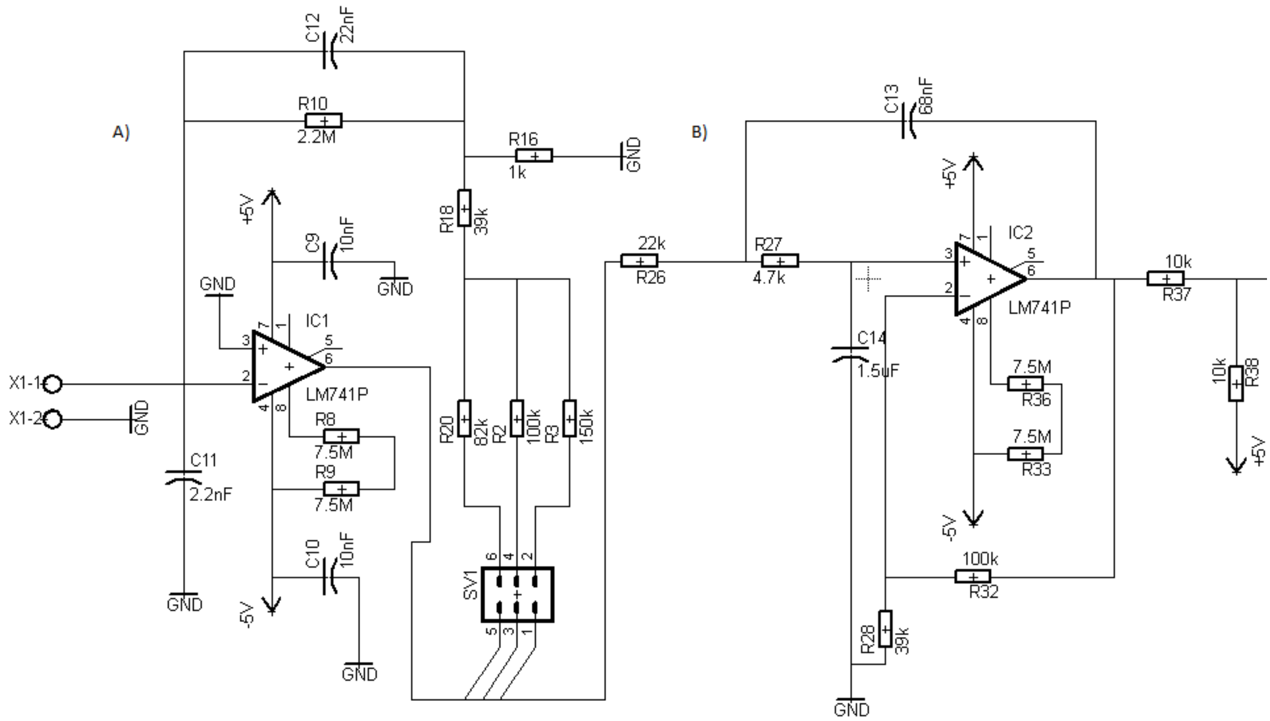

Fig. 4: Signal conditioning circuit. A) charge amplifier stage with selectable gains, B) low pass filter stage.

Transform. It uses a lowpass and highpass special filters to decompose the signal in different frequency bands, then uses decimation to eliminate redundant data and a reconstruct method to obtain the approximation and the details of the waveform. The levels of detail are determined by the user, in this case we used 8 levels of details, and a ‘symlet 6’ mother wavelet. The approximation contains the respiratory signal, and the details 5 and 6 contains the BCG signal. Fig. 5 shows the result of separation algorithm.

The light sensor was calibrated using a standard lux meter, testing different light intensities up to 3000 lux, as shown in Fig. 6. A polynomial model is fitted to transform sensor output to lux level.

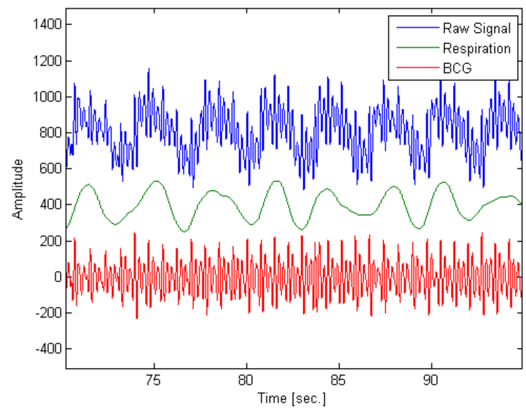

Fig. 5: Wavelet signal processing, the respiration signal correspond to the approximation of the original signal, and the BCG correspond to the details.

Results

We measured people in 2 different surfaces: regular office chairs and a house couch (Fig. 7). The acquired signal is separated in respiration and BCG. To validate this subsystem, a 3-lead ECG is recorded simultaneously. The ECG is preprocessed [6] to increase the QRS amplitude, and the BCG signal was squared to increase the I-J peaks. A simple peak detection algorithm is applied to calculate the beat period (Fig. 8). Pearson correlation corroborates the relationship with $p < 0.001$ and $r = 0.999$.

Ranges of the sound intensity of snoring were established according to the risk of apnea associated with it. Up to 50 [dB] is considered a low risk for apnea, between 50 and 70 [dB] is considered high risk for apnea, and over 70 [dB] the person awakens [7].

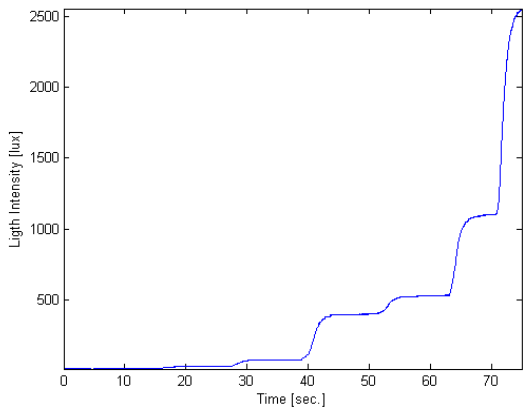

Fig. 6: Light intensity while turning on different sources of light.

Fig. 7: System set-up in a couch. EMFi sensors are 30 cm x 30 cm and covered in foam for protection.

Discussion

The piezoelectric sensors capture physiological signals of the patient unobtrusively, and they are useful to obtain relevant heart and respiration rate from common household furniture. However, our preliminary tests in beds did not produce data as good as in our previous studies [3]. Smaller force sensing resistors are better for beds than larger EMFi, particularly to detect respiration.

The system performed better in a standard couch (Fig. 7) than in a normal office chair. The chair had a small backrest and a harder surface, and the sensor in that position captures a poor respiration signal. In the couch, the backrest is taller, so the EMFi can be placed higher. The sensor captures a better respiration signal, but the BCG amplitude is diminished. In both cases, the BCG from the seat sensor is clearer. That is because the normal component of the force is stronger in the seat position.

The proposed system is also able to include important ambient conditions such as noise, luminosity and temperature and humidity, particularly relevant for sleep. According to the National Sleep Foundation [8], noise above 40 dB can disrupt or prevent sleep. Intensity of the light is also important, above 100 lux people's alertness increases [9] and it is more difficult to fall asleep. Temperatures above 23 °C and below 12 °C can affect the quality of sleep.

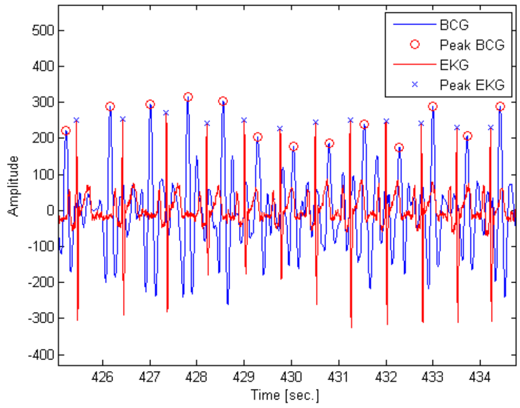

Fig. 8: EKG and BCG measurements and their corresponding peaks.

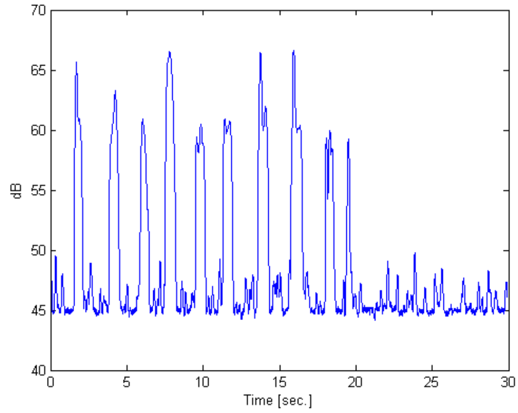

Fig. 9: Simulated snoring during sleep. The intensity of each snoring is useful to complement the respiration signal and determinate if an apnea occurs.

Fig. 9 shows measured sound while simulating snoring. Ambient noise and snoring are easily detectable with standard sensors. This is useful because the snoring intensity and its periodicity are correlated with apneas [10], and can be used as a secondary source for apnea detection.

Light exposure also affects sleep through the melatonin hormone. When light intensity is high, melatonin is suppressed [11] and causes sleep latency problems [12]. Light exposure is also associated with alertness and wakefulness states. In bright environments, over 100 lux [9], alertness increases.

Temperature and humidity are a key factor that can disrupt sleep quality. In hot and humid environments, the body's ability to sweat is greatly diminished, affecting thermoregulation of the body [13].

Conclusion

The proposed unobtrusive system is a convenient tool for ubiquitous health monitoring. Besides its potential use as a home monitoring device for chronic patients, it can be used in nursing homes and rural or low-complexity health centers as a screening device. The large amount of data captured both at day and night can be used for trending, as relevant information for diagnosis (similar to holter data) and for therapy evaluation and control.

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